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PATENT APPLICATION

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	:	Examiner: Allen C. Ho
KAZUAKI TASHIRO, et al.	)	
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Application No.: 10/695,914	)	
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Filed: October 30, 2003	)	
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For: RADIATION IMAGING APPARATUS AND RADIATION IMAGING SYSTEM USING THE SAME	) :) :	September 26, 2005

Mail Stop Amendment
Commissioner for Patents
P.O. Box 1450
Alexandria, VA 22313-1450

SUBMISSION OF SWORN TRANSLATION OF PRIORITY DOCUMENT

Sir:

Further to the Amendment dated June 27, 2005, Applicants submit herewith a sworn translation of priority application Japan 2001-132349, filed on April 27, 2001. In accordance with MPEP § 201.15, the Examiner should confirm that Applicants are entitled to the April 27, 2001 priority date. Once the priority date is confirmed, Applicants respectfully request the Examiner remove U.S. Patent No. 6,800,857 (Kajiwara) as a reference against each of the rejected claims supported by the sworn translation.

Applicants' undersigned attorney may be reached in our Costa Mesa, California office at (714) 540-8700. All correspondence should continue to be directed to our below-listed address.

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DECLARATION

I, Hisako ITO, a subject of Japan residing at 4-35-13, Takadanobaba, Shinjuku-ku, Tokyo, 169-0075 Japan, solemnly and sincerely declare:

That I have thorough knowledge of Japanese and English languages; and

That the attached pages contain a correct translation into English of the specification of the following Japanese Patent Application:

APPLICATION NUMBER	DATE OF APPLICATION
2001-132349	April 27, 2001

I hereby declare that all statements made herein of my own knowledge are true, and that all statements made on information and belief are believed to be true; and further, that these statements are made with the knowledge that willful false statements and the like so made, are punishable by fine or imprisonment, or both, under Section 1001, Title 18 of the United States Code, and that such willful false statements may jeopardize the validity of the application or any patent issuing thereon.

Signed this 2nd day of September, 2005

Hisako Ito

Hisako ITO

132349/2001

[Name of Document] Application for Patent

[Reference No.] 4329002

[Date of Filing] April 27, 2001

[Addressee] Commissioner of the Patent Office

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[Title of the Invention] RADIATION IMAGING APPARATUS AND RADIATION IMAGING SYSTEM USING THE SAME

[Number of Claims] 15

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[List of Documents Attached]

[Name of Document] Specification 1

[Name of Document] Drawings 1

[Name of Document] Abstract 1

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[Name of Document] SPECIFICATION

[Title of the Invention] RADIATION IMAGING APPARATUS AND RADIATION IMAGING SYSTEM USING THE SAME

[Claims]

[Claim 1] A radiation imaging apparatus for forming an image by dividing a subject into a plurality of regions, the apparatus comprising:

a scintillator for converting radiation into light, and a plurality of imaging elements, the plurality of imaging elements being bonded together on a base; and

a planarizing layer for forming a common plane between the imaging elements and the scintillator.

[Claim 2] The radiation imaging apparatus according to claim 1, wherein each of the imaging elements comprises a plurality of pixels, and a scanning circuit for reading signals from the pixels and an external terminal part for external connection are formed in a region in which the pixels are formed.

[Claim 3] The radiation imaging apparatus according to claim 2, wherein the external terminal part is formed in a recessed region.

[Claim 4] The radiation imaging apparatus according to claim 3, wherein the depth of the recessed region is the same as or larger than the thickness of the external terminal part.

[Claim 5] The radiation imaging apparatus according any one of claims 1 to 4, wherein the scintillator is composed of cesium iodide (CsI).

[Claim 6] A radiation imaging apparatus for forming an image by dividing a subject into a plurality of regions, the apparatus comprising:

a scintillator for converting radiation into light, and a plurality of imaging elements; the imaging elements each comprising a plurality of pixels and being bonded together on a base;

a first planarizing layer formed on a pixel formation surface; and

a second planarizing layer for forming a common plane between the imaging elements and the scintillator.

[Claim 7] The radiation imaging apparatus according to claim 6, wherein a scanning circuit for reading signals from the pixels and an external terminal part for external connection are formed in a region in which the pixels of each imaging element are formed, and the first planarizing layer is formed avoiding the external terminal part.

[Claim 8] The radiation imaging apparatus according to claim 7, wherein the thickness of the first planarizing layer is the same as or larger than the thickness of the external terminal part.

[Claim 9] The radiation imaging apparatus according any

one of claims 6 to 8, wherein the scintillator is composed of cesium iodide (CsI).

[Claim 10] A radiation imaging apparatus for forming an image by dividing a subject into a plurality of regions, the apparatus comprising:

a scintillator for converting radiation into light, and a plurality of imaging elements; the imaging elements each comprising a plurality of pixels and being bonded together on a base;

wherein an external terminal part for external connection is formed at the boundary between the adjacent imaging elements, and a step is formed between a region where the external terminal part is formed and a pixel formation region so that the external terminal part is formed at the bottom of the step.

[Claim 11] The radiation imaging apparatus according to claim 10, wherein the depth of the step on which the external terminal part is formed is the same as or larger than the thickness of the external terminal part.

[Claim 12] The radiation imaging apparatus according claim 10 or 11, wherein a planarizing layer is formed on a surface, on which the pixels of the imaging elements are formed, to cover the imaging elements, and the scintillator is formed on the planarizing layer.

[Claim 13] The radiation imaging apparatus according

claim 10 or 11, wherein a planarizing layer is formed on a surface, on which the pixels of the imaging elements are formed, to cover the imaging elements, and a light guide member is provided on the planarizing layer, for guiding light converted by the scintillator to the imaging elements.

[Claim 14] The radiation imaging apparatus according to any one of claims 10 to 13, wherein the scintillator is composed of cesium iodide (CsI).

[Claim 15] A radiation imaging system comprising a radiation source for irradiating a subject with radiation, and the radiation imaging apparatus according to any one of claims 1 to 14, so that the radiation transmitted through the subject is detected by the radiation imaging apparatus to take a radiation image.

[Detailed Description of the Invention]

[0001]

[Technical Field of the Invention]

The present invention relates to a radiation imaging apparatus and a radiation imaging system for reading subject images using high-energy radiation such as X-rays, γ-rays, and the like.

[0002]

[Description of the Related Art]

In various medical fields, digitization of information has recently been advanced. In the field of X-ray diagnosis,

a two-dimensional imaging apparatus has been developed for digitizing image information. For example, an imaging apparatus of a forty-three-cm square at most has been developed for use in mammography and chest radiography.

[0003]

Such an imaging apparatus comprises a plurality of imaging elements which are arranged in a tile-like form to realize a large-area X-ray imaging apparatus. The imaging elements include CCD imaging elements, MOS imaging elements, CMOS imaging elements, and the like. Conventional techniques for this type of apparatus are disclosed in US Patent Nos. 4,810,881 and 5,059,800.

[0004]

U.S. Patent No. 4,810,881 discloses an example of an X-ray imaging panel in which a plurality of modules each comprising a glass substrate and a member of a fluorescent material or the like formed on the glass substrate are connected. Fig. 13 is a sectional view of the imaging panel, taken along the arrangement direction of the modules.

In Fig. 13, reference numeral 3 denotes a reading means; reference numeral 4, a module; reference numeral 5, a glass substrate; reference numeral 6, an X-ray shield; reference numeral 8, a connection line for connecting a line connection portion 30 and the reading means 3; reference numeral 30, the line connection portion; reference numeral

80, a transparent conductive layer; reference numeral 90, a photo-detection layer; and reference numeral 107, a scintillator. A combination of a plurality of these modules provides a large-area X-ray imaging apparatus.

[0005]

Fig. 14 is a sectional view showing the X-ray imaging apparatus disclosed in US Patent No. 5,059,800, taken along the arrangement direction of scintillators. In Fig. 14, reference numeral 14 denotes X-rays, reference numeral 101 denotes an imaging element, reference numerals 20 and 22 each denote a groove formed in a scintillator, reference numeral 107 denotes the scintillator, reference numerals 24 and 25 each denote a reflector filling in the grooves, reference numeral 26 denotes a pixel region of each imaging element, reference numeral 28 denotes an amplifier or a connection region of each imaging element, and reference numeral 90 and 92 each denote a lead wire. Each of the scintillators is formed avoiding the amplifier or the connection region of each imaging element so that the width of the scintillator decreases in the direction closer to the corresponding imaging element.

[0006]

In each of the X-ray imaging apparatuses, X-rays are incident on the side on which the scintillators are formed, and thus X-rays are converted to light by the scintillators

and detected by imaging portions such as the photo-detection layers, the imaging elements, or the like.

[0007]

[Problems to be Solved by the Invention]

However, the above-described conventional techniques have the following problems: In the conventional technique shown in Fig. 13, the scintillator is uniformly formed on each of the modules. However, when the scintillators are formed after the modules are combined, the scintillators cannot be satisfactorily formed in the spaces between the adjacent modules because spaces are present in the imaging regions between the respective modules or between fluorescent materials. Therefore, nonuniformity occurs in the scintillators near the spaces, and a uniform light quantity distribution cannot be obtained. In particular, when the scintillators are formed by vapor deposition of CsI(TI) or CsI(Na), a crystal is abnormally grown in step portions between the modules, thereby causing difficulty in uniformly forming the scintillators over the whole of the imaging apparatus.

[0008]

In the conventional technique shown in Fig. 14, when a plurality of the imaging elements is bonded together on the same substrate, the non-imaging regions 28 each comprising the amplifier or the connection portion are formed, and the

lead wires 90 for electrically connecting the adjacent imaging elements 101 project above the imaging elements. Therefore, when the scintillators are formed after the imaging elements 101 are bonded together, the scintillators must be formed avoiding the non-imaging regions 28 and the lead wires 90 to produce dead spaces in these portions. Consequently, pixel defects occur, or high-resolution imaging cannot be performed.

[0009]

In consideration of the above-described problems, an object of the present invention is to provide a radiation imaging apparatus and a radiation imaging system which are capable of providing images without a joint or defect in realizing an imaging apparatus using a plurality of imaging elements.

[0010]

[Means for Solving the Problems]

In order to achieve the object, the present invention has the following constitution:

1. A radiation imaging apparatus for forming an image by dividing a subject into a plurality of regions, the apparatus comprising a scintillator for converting radiation into light, and a plurality of imaging elements which are bonded together on a base, and a planarizing layer forming a common plane between the imaging elements and the

scintillator.

[0011]

2. The radiation imaging apparatus described above in item 1, wherein each of the imaging elements comprises a plurality of pixels, and a scanning circuit for reading signals from the pixels and an external terminal part for external connection are formed in a region in which the pixels are formed.

[0012]

3. The radiation imaging apparatus described above in item 2, wherein the external terminal part is formed in a recessed region.

[0013]

4. The radiation imaging apparatus described above in item 3, wherein the depth of the recessed region is the same as or larger than the thickness of the external terminal part.

[0014]

5. The radiation imaging apparatus described above in any one of items 1 to 4, wherein the scintillator is composed of cesium iodide (CsI).

[0015]

6. A radiation imaging apparatus for forming an image by dividing a subject into a plurality of regions, the apparatus comprising a scintillator for converting radiation

into light, and a plurality of imaging elements, the imaging elements each comprising a plurality of pixels and being bonded together on a base, a first planarizing layer formed on a pixel formation surface, and a second planarizing layer for forming a common plane between the imaging elements and the scintillator.

[0016]

7. The radiation imaging apparatus described above in item 6, wherein a scanning circuit for reading signals from the pixels and an external terminal part for external connection are formed in a region in which the pixels of each imaging element are formed, and the first planarizing layer is formed avoiding the external terminal part.

[0017]

8. The radiation imaging apparatus described above in item 7, wherein the thickness of the first planarizing layer is the same as or larger than the thickness of the external terminal part.

[0018]

9. The radiation imaging apparatus described above in any one of items 6 to 8, wherein the scintillator is composed of cesium iodide (CsI).

[0019]

10. A radiation imaging apparatus for forming an image by dividing a subject into a plurality of regions, the

apparatus comprising a scintillator for converting radiation into light, and a plurality of imaging elements, the imaging elements each comprising a plurality of pixels and being bonded together on a base, wherein an external terminal part for external connection is formed at the boundary between the adjacent imaging elements, and a step is formed between a region where the external terminal part is formed and a pixel formation region so that the external terminal part is formed at the bottom of the step.

[0020]

11. The radiation imaging apparatus described above in item 10, wherein the depth of the step on which the external terminal part is formed is the same as or larger than the thickness of the external terminal part.

[0021]

12. The radiation imaging apparatus described above in item 10 or 11, wherein a planarizing layer is formed on a surface, on which the pixels of the imaging elements are formed, to cover the imaging elements, and the scintillator is formed on the planarizing layer.

[0022]

13. The radiation imaging apparatus described above in item 10 or 11, wherein a planarizing layer is formed on a surface, on which the pixels of the imaging elements are formed, to cover the imaging elements, and a light guide

member is provided on the planarizing layer, for guiding light converted by the scintillator to the imaging elements.

[0023]

14. The radiation imaging apparatus described above in any one of items 10 to 13, wherein the scintillator is composed of cesium iodide (CsI).

[0024]

15. A radiation imaging system comprising a radiation source for irradiating a subject with radiation, and the radiation imaging apparatus described above in any one of items 1 to 14, so that the radiation transmitted through the subject is detected by the radiation imaging apparatus to take a radiation image.

[0025]

[Description of the Embodiments]

Embodiments of the present invention will be described in detail below with reference to the attached drawings. In each of the embodiments, X-rays are used as incident radiation, but α-rays, β-rays, γ-rays, or the like can be used.

[0026]

(First Embodiment)

Fig. 1 is a plan view showing the constitution of a first embodiment of the present invention. In Fig. 1, nine 138-mm square imaging elements are arranged in a tile-like

form to form a 414-mm square imaging area. Each of the imaging elements 101 comprises a plurality of pixels which are arranged in a two-dimensional form over the entire surface to form a pixel formation area over the entire surface. Namely, the entire surface of each imaging element severs as an imaging area. The imaging elements 101 is disposed on a base 109.

[0027]

Furthermore, a planarizing layer 106, a scintillator 107 and a reflector plate 108 are formed to cover the imaging elements 101. The nine imaging elements 101 have the same shape and the same electric characteristics.

[0028]

Fig. 2 is a plan view showing an eight-inch wafer. In this embodiment, each of the 138-mm square CMOS-type imaging elements 101 is cut out of the wafer. Each of the imaging elements 101 is cut out of such a wafer.

Also, a vertical shift register 203 serving as a vertical scanning circuit, which is a scanning circuit for reading out signals from pixels, and a horizontal shift register 202 similarly serving as a horizontal scanning circuit, which is a scanning circuit for reading out signals from the pixels, are formed in each of the imaging elements 101. In Fig. 2, reference numeral 102 denotes an electrode pad (external terminal part) formed at an end of each

imaging element 101.

[0029]

In this embodiment, the vertical shift register 203 and the horizontal shift register 202 are formed within the pixel formation area of each of the imaging elements 101, and thus no dead space occurs in the peripheries of the imaging elements 101. Therefore, by arranging the imaging elements in a tile-like form, a radiation imaging apparatus having a continuous imaging area can be constructed.

[0030]

Fig. 3 is a sectional view taken along line A-A in Fig. 1. In Fig. 3, reference numeral 101 denotes the imaging elements; reference numeral 104; a flexible substrate serving as a lead for electrically connecting each electrode pad 102 and an external processing substrate 110; reference numeral 106, a planarizing layer formed to cover the imaging elements 101; and reference numeral 107, the scintillator (fluorescent material) for converting incident radiation into light. In this embodiment, the scintillator 107 is formed on the planarizing layer 106. The type of scintillator 107 is preferably selected so that the light emission wavelength is suitable for the sensitivity of the imaging elements 101. Reference numeral 108 denotes a reflection plate; reference numeral 109, the base on which the imaging elements 101 are arranged; reference numeral 110,

the external processing substrate having a circuit for supplying a power supply, a clock, and the like to the imaging elements 101, and for taking out and processing signals from the imaging elements 101; and reference numeral 111, incident X-rays.

[0031]

When the X-rays 111 are incident, as shown in Fig. 3, the X-rays 111 are converted into light by the scintillator 107, and the light is absorbed and converted into electrical signals by the photodiodes of the pixels arranged in a two-dimensional form in each imaging element 101. The electrical signals are processed by the external processing substrate 110 through the flexible substrates 104.

[0032]

Fig. 4 is an enlarged view of region B shown by a circle in Fig. 1. In Fig. 4, reference numeral 102 denotes the electrode pad formed in one pixel region at an end of each imaging element 101; reference numeral 103, a bump formed on each of the electrode pads 104; reference numeral 104, the flexible substrate; and reference numeral 112, a pixel including an element for converting light into an electrical charge. A connection portion for connecting each flexible substrate 104 is formed a one-step recess 121.

[0033]

As shown in Fig. 4, in each of the imaging elements 101,

160-μm square pixels 112 are formed in a two-dimensional form over the entire surface of the imaging element 101, the adjacent imaging elements 101 being arranged with no space therebetween.

[0034]

The spaces between the respective imaging elements 101 are about 50 μm which is sufficiently smaller than the pixel pitch of 160 μm. Each of the flexible substrates 104 is extended through the space between the adjacent imaging elements 101. The flexible substrates 104 are formed to a thickness of as small as about 30 μm.

[0035]

Therefore, the spaces between the respective imaging elements 101, after the imaging elements 101 are bonded together, can be set to about 50 μm, and no dead space is present in the peripheries of the imaging elements 101 to form a state in which no defect occurs in a read image, i.e., a state with substantially no space. In this embodiment, the external terminal part 120 of each imaging element 101 is formed in a region one-step lower than the pixel formation region.

[0036]

Figs. 5 and 6 are sectional views showing region C shown in Fig. 3. The configuration of the region C will be described with reference to the process for forming the

external terminal part 120. First, the electrode pad 102 is formed in a pixel region at an end of each of the imaging elements 101. This pixel region is formed in a region one-step lower than the pixel formation region, i.e., formed in a concave shape (recess 121), to form step S. Each of the electrode pads 102 is provided in the one-step recessed portion, the bump 103 is formed thereon, and the flexible substrate 104 is bent along the edge of each imaging element 101 and connected to the bump 103.

[0037]

The nine imaging elements 101 each having the external terminal part 120 formed as described above are arranged so that the pixel formation surfaces are at the same height, and then bonded in a tile-like form on the base 109. Then, the flexible substrate 104 extending from each of the imaging elements 101 is electrically connected to the external processing substrate 110 disposed as shown in Fig. 3.

[0038]

The external terminal parts 120 are formed in recessed regions lower than the pixel formation region, and the depth of the recesses 121 is larger than the thickness of the external terminal parts 120. The depth may be the same as the thickness of the external terminal parts 120. In this embodiment, the depth is 45 μm, and the thickness of the

external terminal parts 120 is about 40 μm. As shown in Fig. 6, a deviation of about 5 μm may occur between the respective imaging elements.

[0039]

Furthermore, an adhesive 105 is applied to the imaging elements 101, which are bonded together, so as to fill in the recesses 121 and the spaces between the adjacent imaging elements 101. Then, a planarizing member serving as the planarizing layer 106 is formed thereon. As the planarizing layer 106, a lead-containing X-ray shielding glass plate having a thickness of 50 μm is used.

[0040]

As described above, the planarizing layer 106 is formed on the imaging elements 101 to planarize the pixel formation surfaces of the imaging elements 101, the external terminal parts 120, and the spaces between the respective imaging elements 101 at the same height over all substrate surfaces.

[0041]

Even when a deviation occurs between the pixel formation surfaces of the imaging elements 101, as shown in Fig. 6, the planarizing layer 106 forms a planarized surface above the imaging element 101 at the same height and absorbs a deviation between the substrates, thereby forming a plane above the imaging elements 101.

[0042]

By forming the planarizing layer 106, steps and deviations can be absorbed to provide a common uniform plane for the scintillator formed thereon.

[0043]

Then, fluorescent material CsI is deposited on the plane comprising the planarizing layer 106 by vaporization to form the uniform scintillator 107 above the imaging elements 101.

[0044]

When the scintillator 107 is formed as described above, the scintillator 107 can be uniformly formed without deviation due to abnormal growth of the fluorescent material only in the external terminal parts and the spaces between the respective imaging elements. Therefore, uniform light emission can be achieved, and variation in a light quantity distribution can be decreased.

[0045]

When the distance L between the lower surface of the scintillator 107 and the pixel formation surfaces is large, light from the scintillator 107 diffuses between the scintillator 107 and the pixel formation surfaces to possibly blur an image, thereby deteriorating resolution. However, in this embodiment, each of the external terminal parts 120 is formed in the region lower than the pixel formation surface, and thus the distance L is determined by

the thickness of the planarizing layer 106 and the adhesive 105 regardless of the thickness of the external terminal parts 120. Therefore, the distance L can be decreased to 60 μm or less to prevent deterioration in resolution.

[0046]

As a member used for the planarizing layer 106, a polyimide film having no radiation shielding ability, but easier to handle than the glass plate and having transparency, may be used. Alternatively, a polyimide resin or the like may be uniformly applied and cured to form the planarizing layer 160 without using the adhesive 105.

[0047]

When the scintillator 107 is formed by an evaporation process using cesium iodide (CsI) or by a coating process using a transparent resin in which a powder scintillator made of Gd_2O_2S or the like containing europium or terbium as an activator is dispersed, the same effect as described above can be obtained.

[0048]

Fig. 7 is a drawing showing a configuration of one pixel of the CMOS-type imaging elements 101. In Fig. 7, reference numeral 601 denotes a photodiode for photoelectric conversion; reference numeral 602, a floating diffusion for storing electric charges; reference numeral 603, a transfer MOS transistor (transfer switch) for transferring the

charges produced in the photodiode to the floating diffusion; reference numeral 604, a reset MOS transistor (reset switch) for discharging the charges stored in the floating diffusion; reference numeral 605, a line selection MOS transistor (line selection switch) for selecting a line; and reference numeral 606, an amplification MOS transistor (pixel amplifier) functioning as a source follower.

[0049]

Fig. 8 is a drawing showing equivalent circuits of an imaging element comprising 3×3 pixels using the pixel shown in Fig. 7. Gates of the transfer MOS transistors 603 are connected to ϕTX 707 extended from the vertical shift register 203, gates of the reset MOS transistors 604 are connected to ϕRES 708 extended from the vertical shift register 203, and gates of the line selection MOS transistors 606 are connected to ϕSEL 709 extended from the vertical shift register 203. In Fig. 8, the floating diffusions 602 are omitted.

[0050]

The photoelectric conversion is performed by the photodiodes 601. During the time of storage of photoelectric charge, the transfer MOS transistors 603 are turned off so that charges subjected to photoelectric conversion by the photodiodes 601 are not transferred to the gates of the amplification MOS transistors 606 constituting

the pixel amplifiers. Before storage, the reset MOS transistors 604 are turned on to initialize the gates of the amplification MOS transistors 606 constituting the pixel amplifiers to an appropriate voltage. This corresponds to a dark level.

[0051]

Next, the line selection MOS transistors 605 are turned on to bring the amplification MOS transistors 606 each comprising a load current source 710 and a pixel amplifier 715 in an operating state. Then, the transfer MOS transistors 603 are turned on to transfer the charges stored in the photodiodes 601 to the gates of the amplification MOS transistors 606 constituting the pixel amplifiers.

[0052]

When the output of the selected line is produced on a vertical output line (signal output line) 711, a corresponding column selection switch (multiplexer) 713 is driven by the horizontal shift register 202 to read the output, transferring the read output to the output amplifier 715.

[0053]

Fig. 9 is a drawing showing a state in which a unit block (a unit for selecting and driving one line) 801 of the vertical shift register is arranged in each pixel region (each cell) 803 together with a one-pixel circuit 802. The

one-pixel circuit 802 is the same as shown in Fig. 6. Each of the circuits constituting the vertical shift resistor is a simple circuit comprising a static shift register 804 and a transfer gate 805, for transmitting a transfer signal, a reset signal and a line selection signal to ϕTX 707, ϕRES 708, and ϕSEL 709, respectively. The static shift register 804, and the transfer gate 805 are driven by signals from a clock signal line (not shown in the drawing). The circuit configuration of the shift register is not limited to this, and any desired circuit configuration may be used according to the driving methods for pixel addition, decimation reading, and the like.

[0054]

As shown in Fig. 1, the vertical shift register 203 and the horizontal shift register 202 are arranged in the pixel forming area of each of the imaging elements. Therefore, the unit block 801 of the shift register for processing one line is arranged to be received in the pixel pitch. Namely, these blocks are arranged in a line to form the vertical shift register or the horizontal shift register. The blocks are arranged in a line in each of the vertical direction and the horizontal direction.

[0055]

(Second Embodiment)

A second embodiment of the present invention will be

described in detail below. The planar configuration of the second embodiment is the same as the first embodiment.

[0056]

Fig. 10 is a sectional view showing the second embodiment of the present invention, taken along line A-A in Fig. 1. In the second embodiment, a fiber-optic plate (FOP) 130 of a 1× magnification optical system is disposed as a n X-ray shielding member between the scintillator (fluorescent material) 107 and the imaging elements 101 through a first planarizing layer 106(a) and a second planarizing layer 106(b). X-rays 111 are incident on the scintillator 107 and converted into light.

[0057]

In this embodiment, light transmitted through the fiber-optic plate 111 having a thickness of 3 mm is detected by the imaging elements 101. The fiber-optic plate 130 is of a 1× magnification optical system, and thus the light reaches the imaging elements 101 without blurring. However, X-rays not absorbed by the scintillator 107 are absorbed by the fiber-optic plate 130 and do not reach the imaging elements 101, thereby causing no damage to the imaging elements 101.

[0058]

The type of scintillator 107 is preferably selected so that the light emission wavelength is suitable for the

sensitivity of the imaging elements 111. Reference numeral 110 denotes an external processing substrate having a circuit for supplying a power supply, a clock, and the like to the imaging elements 101, and for taking out and processing signals from the imaging elements 101, and reference numeral 104 denotes a flexible substrate which is a type of wiring for electrically connecting each imaging element 101 and the external processing substrate 110.

[0059]

Fig. 11 is an enlarged view of region C shown by a circle in Fig. 10. The configuration of region C will be described with reference to the process for forming an external terminal part 120. In this embodiment, the external terminal part 120 is disposed in a region of one pixel at an end of each imaging element 101. The first planarizing layer 106(a) is formed for forming step S so that the region of the external terminal part 120 is one-step lower that the peripheral pixel regions.

[0060]

Also, a bump 103 is provided on the external terminal part 120 formed in the one-step lower region, and then, the flexible substrate 104 is bent along the edge portion of each of the imaging elements 101 and connected to the bump 103. Furthermore, nine imaging elements 101 are arranged in a tile-like form so that the pixel surfaces are at the same

height, and the flexible substrates 104 extended from the imaging elements 101 are electrically connected to the external processing substrate 110 arranged at the back of the imaging elements 101.

[0061]

In this state, the total height of the electrode pad 102, the bump 103, and the flexible substrate 104 is about 40 μm. In this embodiment, the first planarizing layer 106(a) is formed on the imaging elements 101 using photosensitive polyimide before the imaging elements 101 bonded in a tile-like form. The photosensitive polyimide is applied to a thickness of 40 μm, exposed to light through a mask provided only for the external terminal parts 120, and then partially removed by etching.

[0062]

As a result, the steps S are formed. The nine imaging elements 101 are bonded in a tile-like form, and the flexible substrates 104 are connected. Then, a first adhesive 105(a) is applied to the bonded imaging elements 101 so that the stepped portions and the spaces between the imaging elements are filled with the first adhesive 105(a). Then, a lead-containing X-ray shielding glass plate having a thickness of 50 μm is disposed as the second planarizing layer 106(b) on the first adhesive 105(a). As a result, the surfaces of the imaging elements, the external terminal

parts 120, and the spaces between the imaging elements are planarized at the same height to form a plane.

[0063]

Then, a second adhesive 105(b) is applied, and a reflector plate 108 and the fiber-optic plate 130 comprising the scintillator 107 composed of CsI(TI) are disposed. The fiber-optic plate 130 does not rise in the regions of the external terminal parts 120 and the spaces between the respective imaging elements 101, and distance L does not significantly varies, thereby causing no variation in a light quantity distribution.

[0064]

In this embodiment, the distance L between the bottom of the scintillator (fluorescent material) and the pixel surfaces is about 100 μm because the first and second planarizing layers 106(a) and 106(b) are used. However, resolution is not significantly decreased. As a planarizing member, a light-transmitting polyimide thin film of about 10 μm in thickness having no radiation shielding ability, but easier to handle than a glass plate and having transparency, may be used. Alternatively, a polyimide resin or the like may be uniformly applied and cured to form the planarizing member without using an adhesive. In this case, the distance L can be further decreased to improve resolution.

[0065]

In this embodiment, the spaces between the respective imaging elements, after the imaging elements are bonded together, can be set to about 50 μm, and substantially no dead space occurs between the imaging elements 101 after the imaging elements 101 are bonded.

[0066]

This embodiment uses the fiber-optic plate with the scintillator (fluorescent material). However, as in the first embodiment, CsI(TI) may be formed on the second planarizing layer by an evaporation process. In this case, distance L can be decreased. The use of a polyimide film or the like as the planarizing layer can further decrease distance L to permit the formation of a thin imaging apparatus, while the X-ray shielding ability is degraded.

[0067]

(Third Embodiment)

Fig. 12 is a drawing showing an example in which a radiation imaging apparatus according to the present invention is applied to an X-ray diagnosis system.

[0068]

X-rays 6060 produced in an X-ray tube 6050 are transmitted through the chest region 6062 of a patient or test subject 6061 and incident on a radiation imaging apparatus 6040 comprising the scintillator 107, FOP 130, the imaging elements 101, and the external processing substrate

110. The incident X-rays contains inside information of the body of the patient 6061.

[0069]

Light is emitted from the scintillator 107 in response to incidence of X-rays, and subjected to photoelectric conversion by the imaging elements 101 to obtain electrical information. The information is converted into a digital signal, and an image is processed by an image processor 6070 according to the digital signal and displayed on a display 6080 in a control room.

[0070]

The information can be transmitted to a remote place by transmission means such as a telephone circuit 6090. The image can be displayed on a display 6081 in a doctor room at a remote location or can be stored in a storage means such as an optical disk or the like. This may permit diagnosis by another doctor at a remote location. Also, the information can be recorded on a film 6110 by a film processor 6100.

[0071]

As described above, according to the present invention, surfaces of imaging elements on which a scintillator is formed is planarized, and thus the scintillator can be uniformly formed, thereby decreasing variation in the quantities of light transmitted through the scintillator and

incident on the imaging elements. When a light guide member is formed, a surface on which the light guide member is formed is also planarized to decrease variations in the quantities of light incident on the imaging elements. Also, pixels are formed in an effective pixel region over the entire surface of each imaging element, and a scanning circuit such as a shift register or the like, an external terminal part, etc. are formed between the pixels in the effective pixel region. Therefore, the imaging elements can be arranged in a tile-like form with substantially no space between the imaging elements. Even when a plurality of imaging elements, for example, five imaging elements (arranged in a cross shape) or nine imaging elements (arranged in a 3×3 matrix form), are arranged for detecting an image, neither discontinuity nor deletion occurs in an image between the imaging elements. Furthermore, an external terminal part is formed in a region lower than the pixel formation region, and thus the distance between the pixels and a scintillator or a light guide member formed on the imaging elements can be decreased to decrease blurring of an image due to an increase in the distance. By utilizing the above-descried configuration, it is possible to use an imaging element made of single crystal silicon, which has difficulty in increasing the size, not made of amorphous silicon. Therefore, a large-screen animation or

high-definition animation with a high S/N ratio can be realized. In addition, the cost can be decreased because tapered FOP need not be used.

[Brief Description of the Drawings]

[Fig. 1]

Fig. 1 is a plan view showing a state in which nine imaging elements 101 are arranged.

[Fig. 2]

Fig. 2 is a plan view showing a wafer for forming an imaging element 101.

[Fig. 3]

Fig. 3 is a sectional view of a first embodiment taken along line A-A in Fig. 1.

[Fig. 4]

Fig. 4 is an enlarged plan view showing region B shown by a circle in Fig. 1.

[Fig. 5]

Fig. 5 is an enlarged sectional view showing region C shown by a circle in Fig. 3 showing the first embodiment.

[Fig. 6]

Fig. 6 is an enlarged sectional view showing region C shown by a circle in Fig. 3 showing the first embodiment.

[Fig. 7]

Fig. 7 is a drawing showing a one-pixel circuit.

[Fig. 8]

Fig. 8 is a drawing showing equivalent circuits of an imaging element.

[Fig. 9]

Fig. 9 is a conceptual plan view showing a case in which a unit block of a shift register is formed adjacent to a corresponding pixel region.

[Fig. 10]

Fig. 10 is a sectional view of a second embodiment of the present invention taken along line A-A in Fig. 2.

[Fig. 11]

Fig. 11 is an enlarged sectional view showing region C shown by a circle in Fig. 10.

[Fig. 12]

Fig. 12 is a conceptual view showing the configuration of a radiation imaging system according to a third embodiment of the present invention.

[Fig. 13]

Fig. 13 is a drawing illustrating conventional technique 1.

[Fig. 14]

Fig. 14 is a drawing illustrating conventional technique 2.

[Reference Numerals]

101 imaging element

102 electrode pad

103 bump

104 flexible substrate

105 adhesive

106 planarizing layer

107 scintillator (fluorescent material)

108 reflector plate

109 base

110 external processing substrate

111 X-rays

112 pixel

120 external terminal part

121 recess

130 fiber-optic plate (FOP)

202 horizontal shift register

203 vertical shift register

6040 radiation imaging apparatus

6050 X-ray tube

6060 X-rays

6061 patient or test subject

6062 chest portion of patient or test subject

6070 image processor

6080 display in control room

6081 display in doctor room

6100 film processor

6110 film

[Name of Document] ABSTRACT

[Abstract]

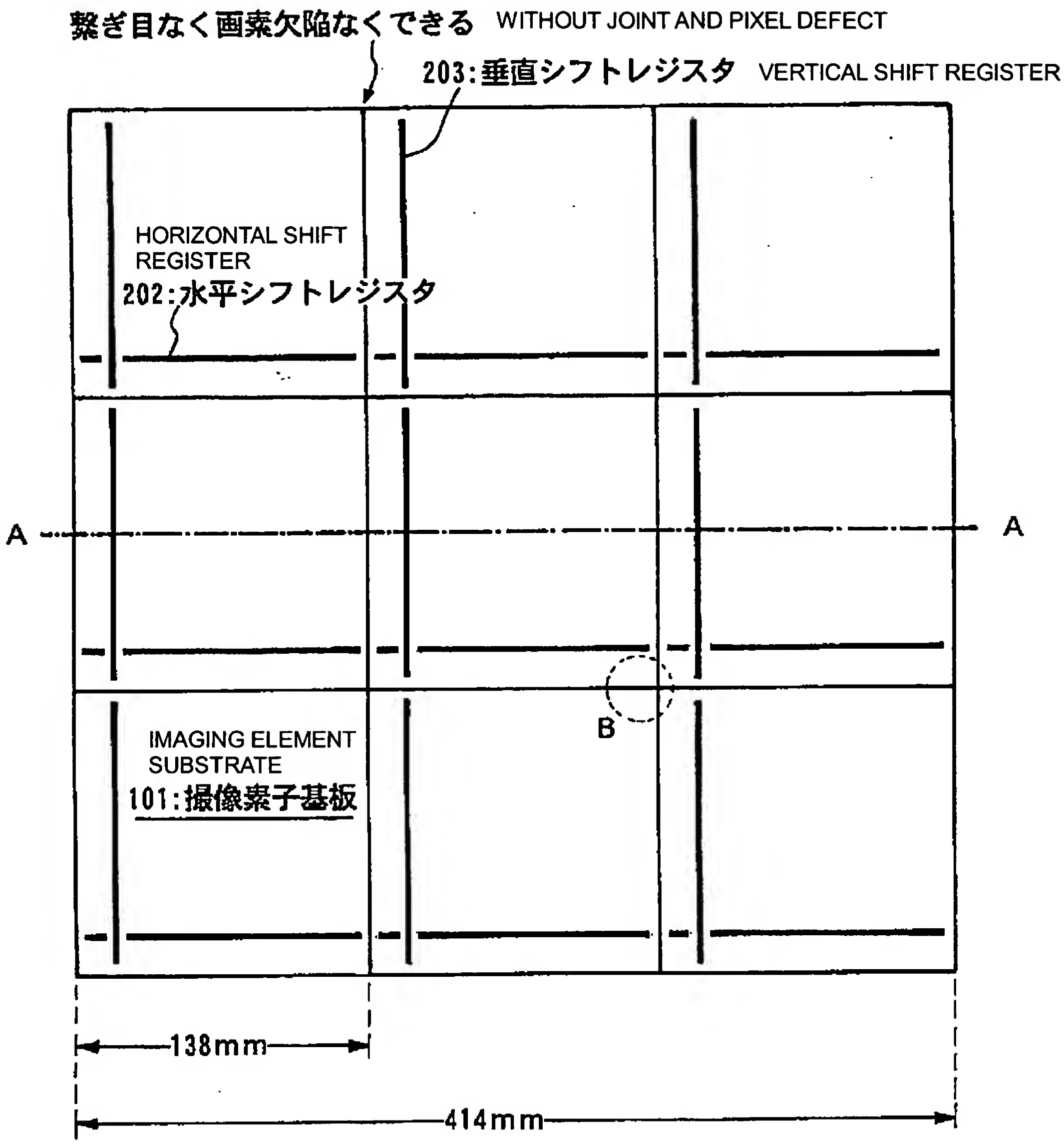
[Object] To provide a radiation imaging apparatus and a radiation imaging system which are capable of providing images without a joint or defect in realizing an imaging apparatus using a plurality of imaging elements.

[Solving Means] A radiation imaging apparatus for forming an image by dividing a subject into a plurality of regions includes a plurality of imaging elements 101 which are bonded together on a base 109, and a planarizing layer 106 for forming a common plane between the imaging elements 101 and a scintillator 107 for converting radiation into light.

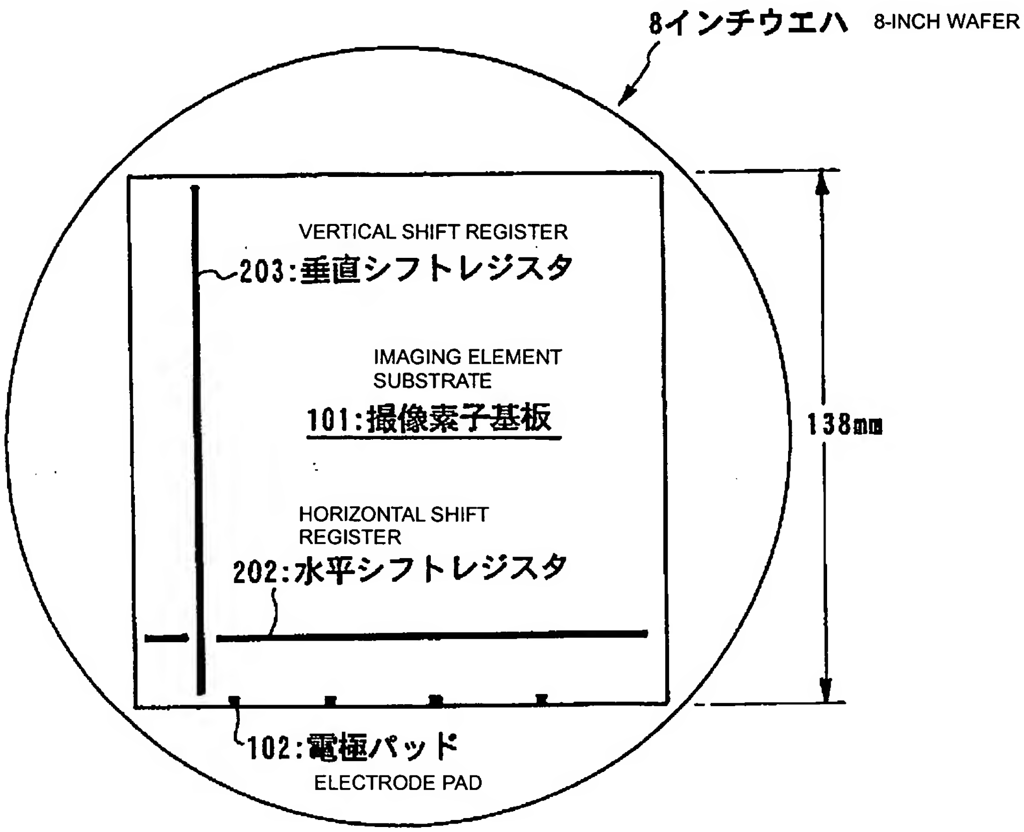
[Selected Figure] Fig. 3

【書類名】　図面　[Name of Document] DRAWINGS

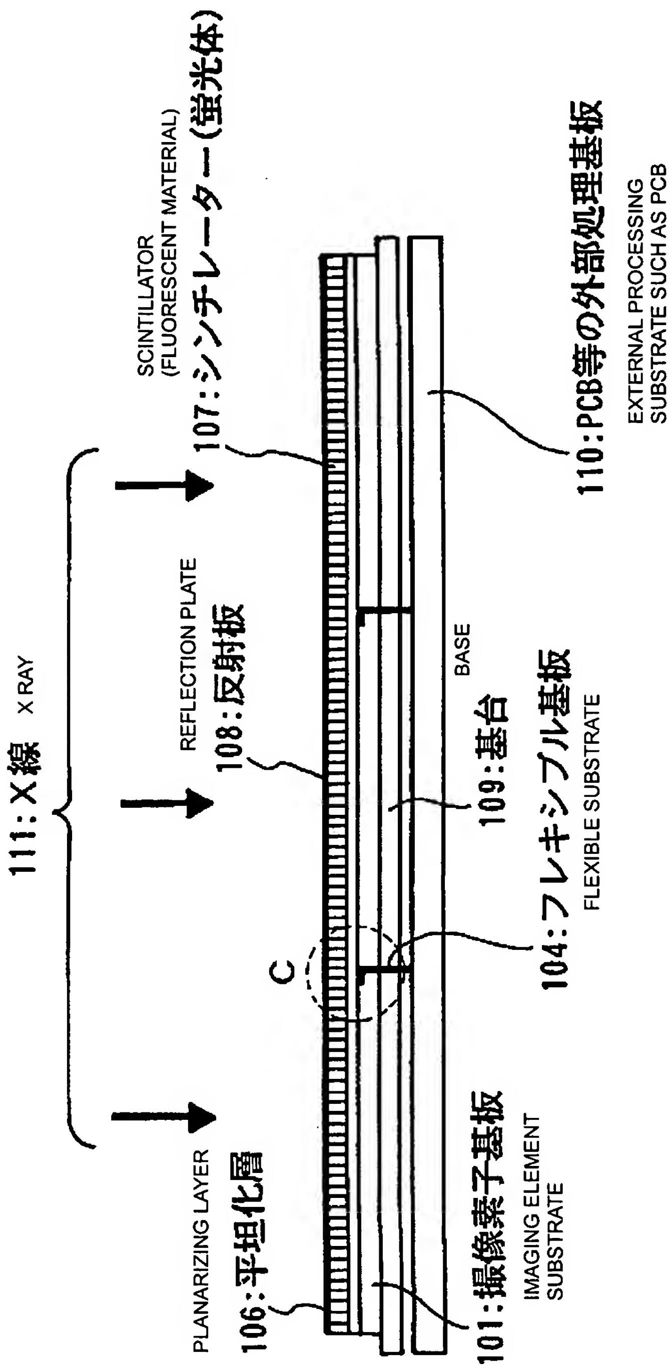
【図１】 [FIG. 1]



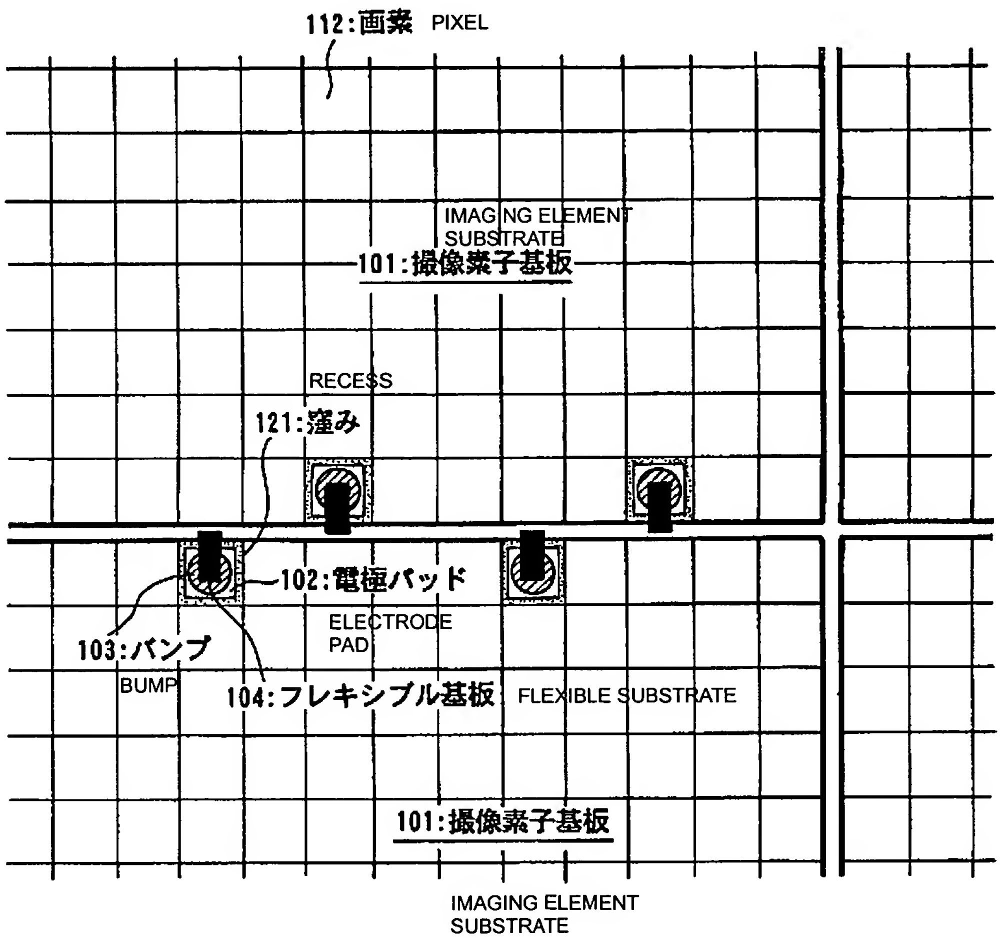
【図２】 [FIG. 2]


8インチウエハ 8-INCH WAFER
VERTICAL SHIFT REGISTER
203:垂直シフトレジスタ
IMAGING ELEMENT SUBSTRATE
101:撮像素子基板
HORIZONTAL SHIFT REGISTER
202:水平シフトレジスタ
138mm
102:電極パッド
ELECTRODE PAD

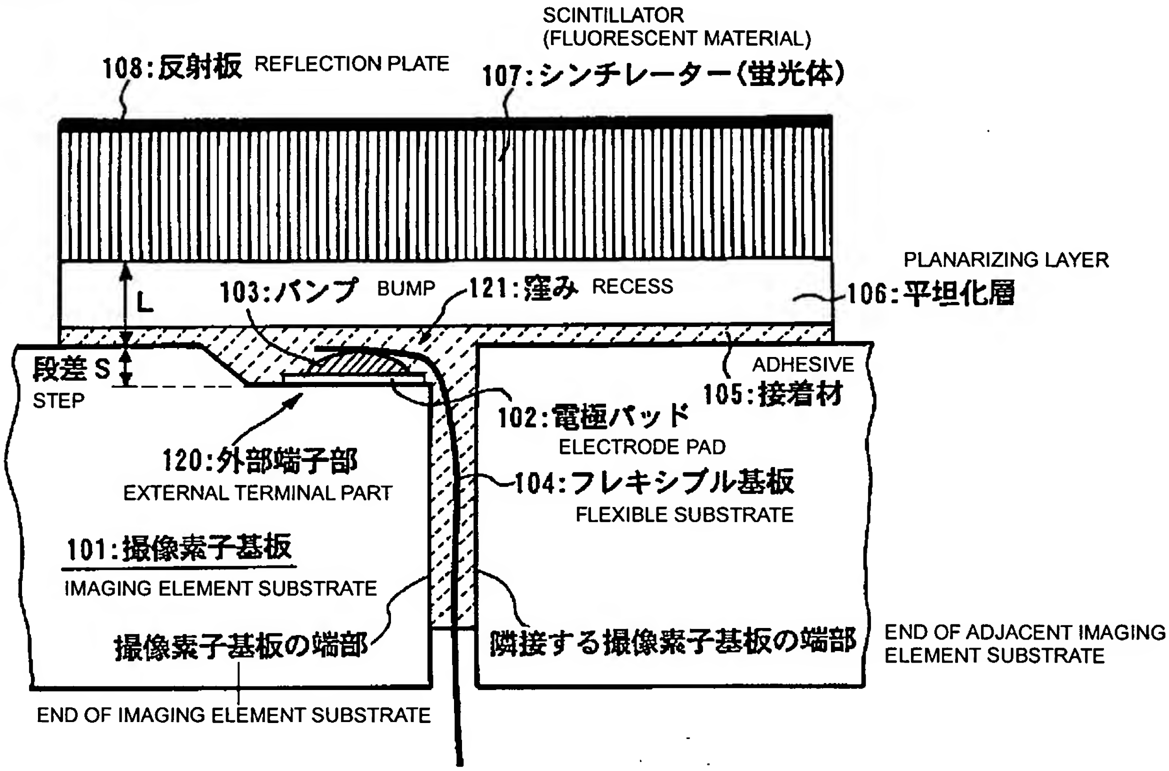
【図３】 [FIG. 3]


111：Ｘ線 X RAY
SCINTILLATOR (FLUORESCENT MATERIAL)
107：シンチレーター（蛍光体）
REFLECTION PLATE
108：反射板
PLANARIZING LAYER
106：平坦化層
C
BASE
109：基台
104：フレキシブル基板
FLEXIBLE SUBSTRATE
101：撮像素子基板
IMAGING ELEMENT SUBSTRATE
110：PCB等の外部処理基板
EXTERNAL PROCESSING SUBSTRATE SUCH AS PCB

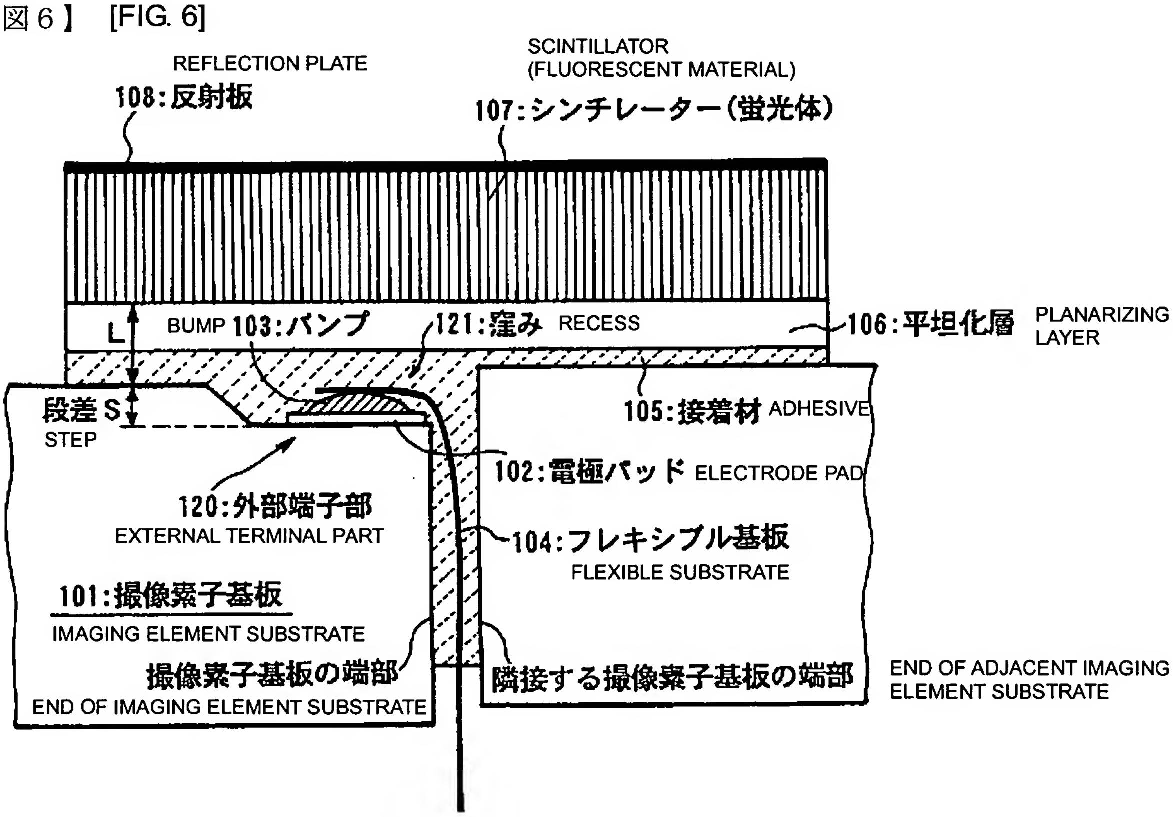
【図４】 [FIG. 4]


112:画素 PIXEL
IMAGING ELEMENT SUBSTRATE
101:撮像素子基板
RECESS
121:窪み
102:電極パッド
ELECTRODE PAD
103:バンプ
BUMP
104:フレキシブル基板
FLEXIBLE SUBSTRATE
101:撮像素子基板
IMAGING ELEMENT SUBSTRATE

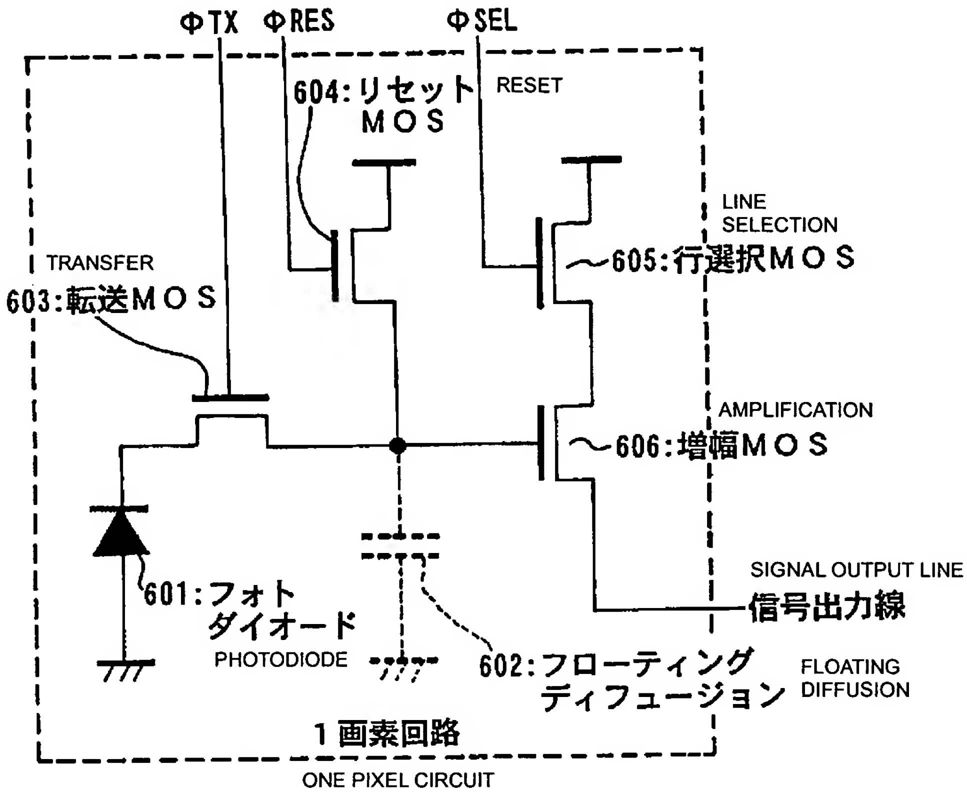
【図５】 [FIG. 5]



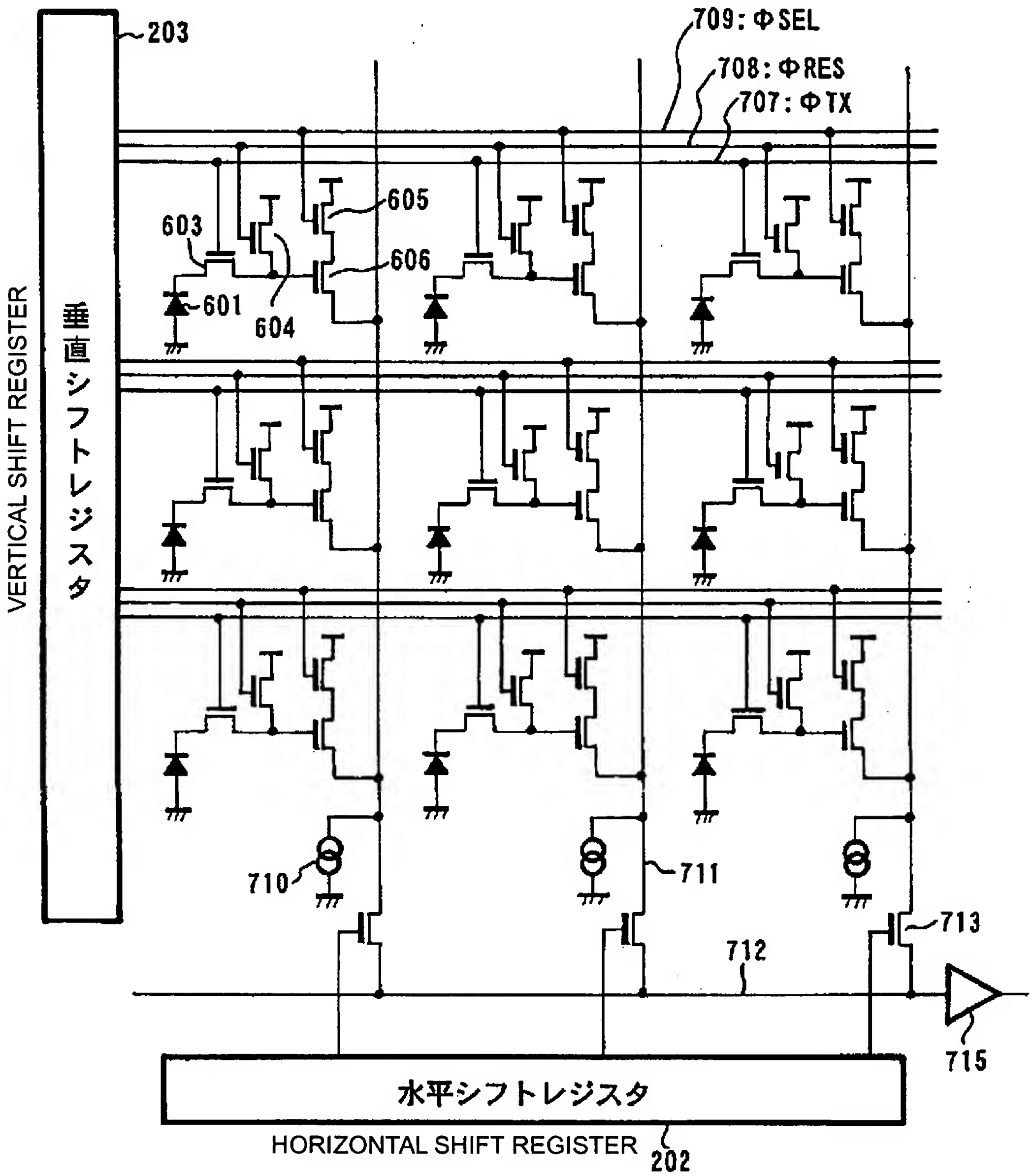
【図６】 [FIG. 6]



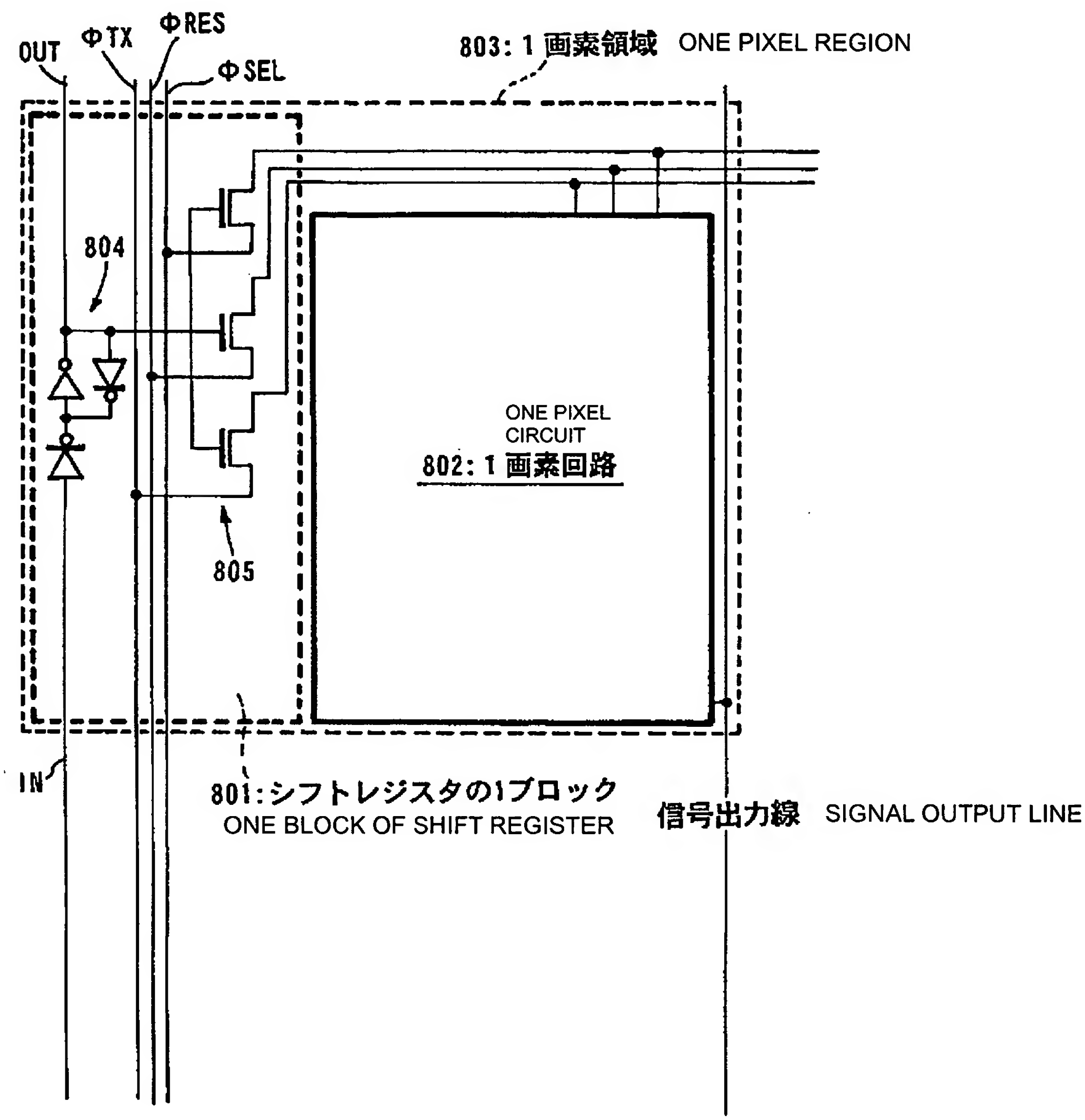
【図７】 [FIG. 7]


ΦTX
ΦRES
ΦSEL
604:リセット
ＭＯＳ
RESET
LINE
SELECTION
605:行選択ＭＯＳ
TRANSFER
603:転送ＭＯＳ
AMPLIFICATION
606:増幅ＭＯＳ
SIGNAL OUTPUT LINE
信号出力線
601:フォト
ダイオード
PHOTODIODE
602:フローティング
ディフュージョン
FLOATING
DIFFUSION
１画素回路
ONE PIXEL CIRCUIT

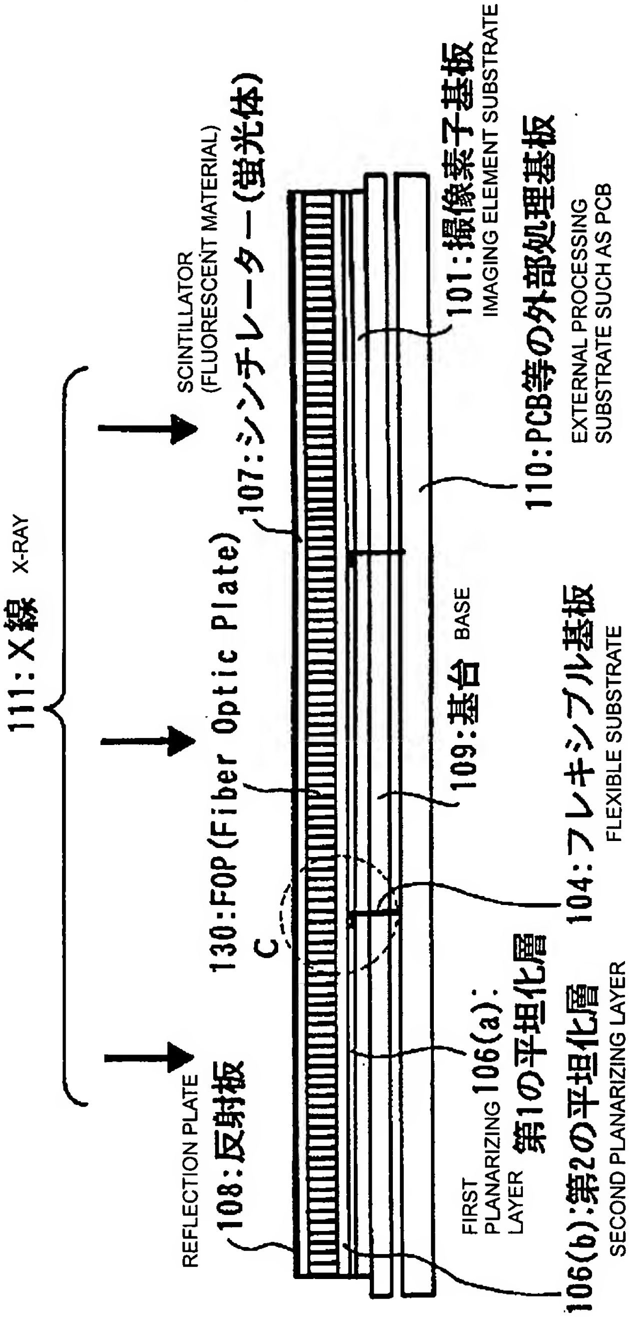
【図８】　[FIG. 8]


709: ΦSEL
708: ΦRES
707: ΦTX
203
605
603
606
601
604
VERTICAL SHIFT REGISTER
垂直シフトレジスタ
710
711
713
712
715
水平シフトレジスタ
HORIZONTAL SHIFT REGISTER
202

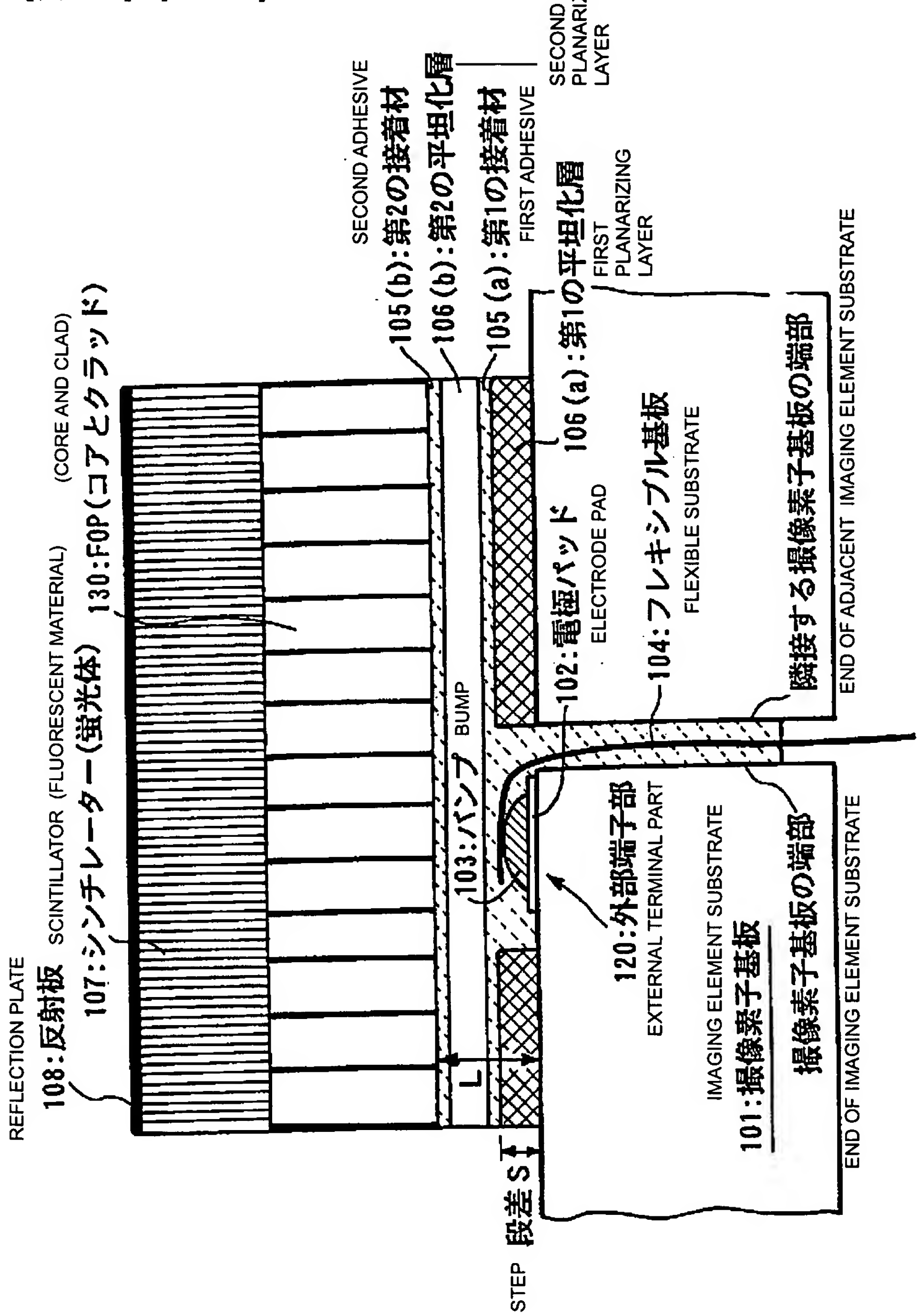
【図９】 [FIG. 9]


OUT
ΦTX
ΦRES
ΦSEL
803: 1 画素領域 ONE PIXEL REGION
804
ONE PIXEL
CIRCUIT
802: 1 画素回路
805
IN
801:シフトレジスタの1ブロック
ONE BLOCK OF SHIFT REGISTER
信号出力線 SIGNAL OUTPUT LINE

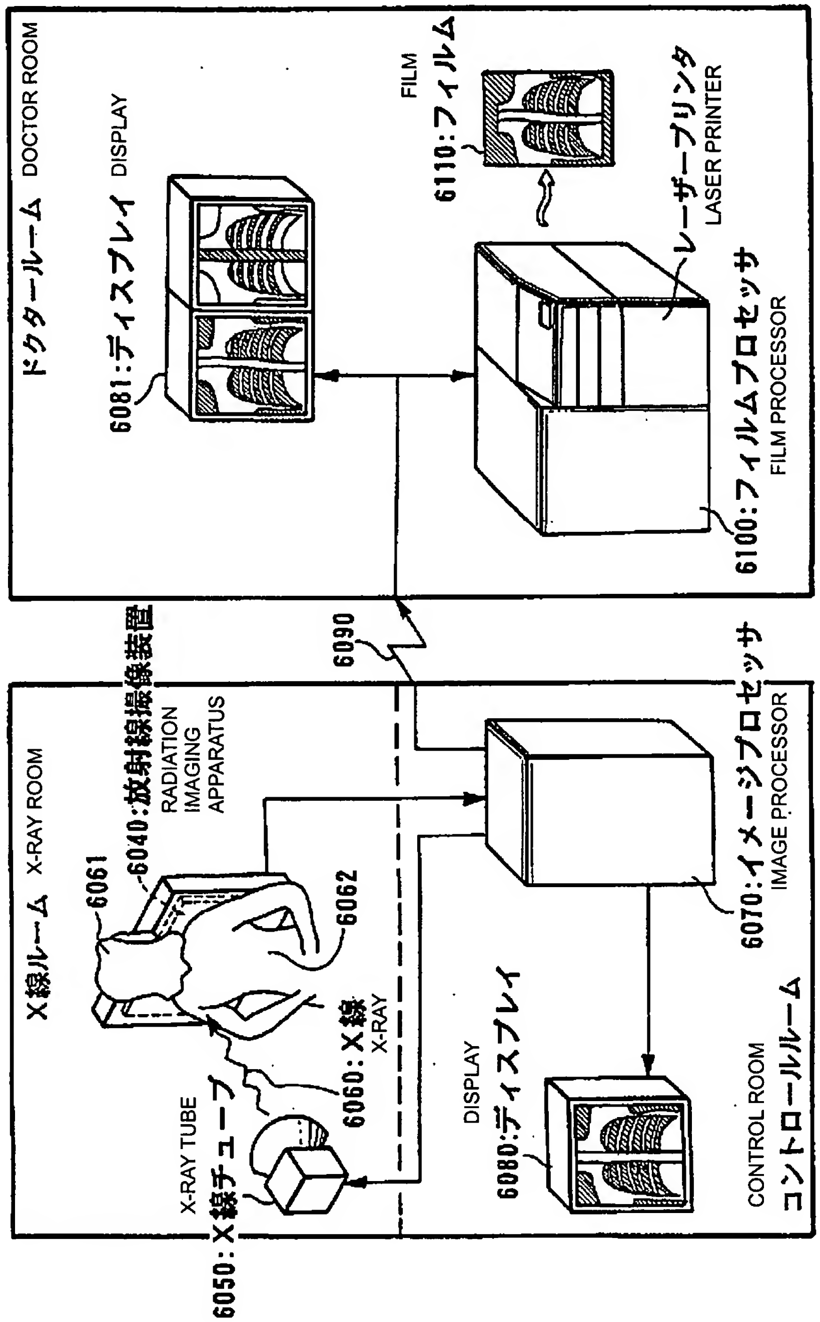
【図１０】 [FIG. 10]


111:X線 X-RAY
107:シンチレーター(蛍光体)
SCINTILLATOR (FLUORESCENT MATERIAL)
130:FOP(Fiber Optic Plate)
108:反射板
REFLECTION PLATE
C
101:撮像素子基板
IMAGING ELEMENT SUBSTRATE
110:PCB等の外部処理基板
EXTERNAL PROCESSING SUBSTRATE SUCH AS PCB
109:基台 BASE
104:フレキシブル基板
FLEXIBLE SUBSTRATE
106(a):第1の平坦化層
FIRST PLANARIZING LAYER
106(b):第2の平坦化層
SECOND PLANARIZING LAYER

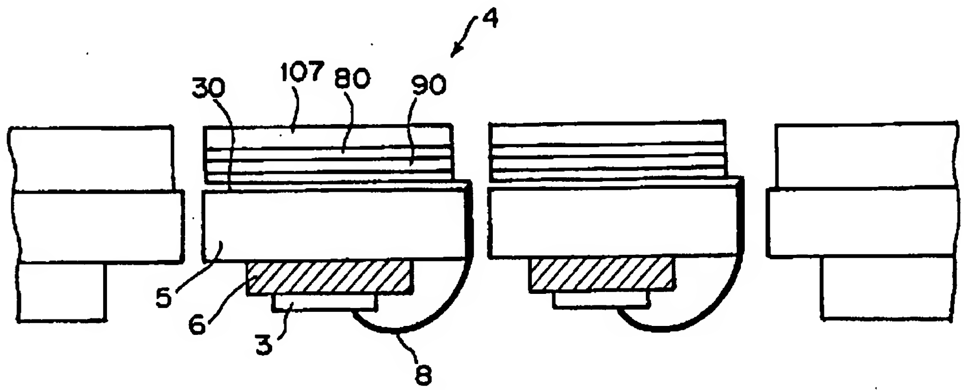
【図１１】 [FIG. 11]


108:反射板
REFLECTION PLATE
107:シンチレーター(蛍光体)
SCINTILLATOR (FLUORESCENT MATERIAL)
130:FOP(コアとクラッド)
(CORE AND CLAD)
105(b):第2の接着材
SECOND ADHESIVE
106(b):第2の平坦化層
SECOND PLANARIZING LAYER
105(a):第1の接着材
FIRST ADHESIVE
106(a):第1の平坦化層
FIRST PLANARIZING LAYER
102:電極パッド
ELECTRODE PAD
104:フレキシブル基板
FLEXIBLE SUBSTRATE
隣接する撮像素子基板の端部
END OF ADJACENT IMAGING ELEMENT SUBSTRATE
103:バンプ
BUMP
120:外部端子部
EXTERNAL TERMINAL PART
101:撮像素子基板
IMAGING ELEMENT SUBSTRATE
撮像素子基板の端部
END OF IMAGING ELEMENT SUBSTRATE
L
段差 S
STEP

【図１２】 [FIG. 12]


ドクタールーム DOCTOR ROOM
6081:ディスプレイ DISPLAY
6110:フィルム FILM
レーザープリンタ LASER PRINTER
6100:フィルムプロセッサ FILM PROCESSOR
6090
X線ルーム X-RAY ROOM
6040:放射線撮像装置 RADIATION IMAGING APPARATUS
6061
6062
6060:X線 X-RAY
6050:X線チューブ X-RAY TUBE
6070:イメージプロセッサ IMAGE PROCESSOR
6080:ディスプレイ DISPLAY
CONTROL ROOM コントロールルーム

【図１３】 [FIG. 13]



【図１４】 [FIG. 14]